COMBINING APICAL AND PARASTERNAL VIEWS TO IMPROVE MOTION ESTIMATION IN REAL-TIME 3D ECHOCARDIOGRAPHIC SEQUENCES

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ABSTRACT

Real-time 3D echocardiography (RT3DE), while offering obvious advantages over traditional two-dimensional echocardiography, is hampered by its limited image quality. This makes it difficult to apply accurate quantitative analysis tools that fully exploit this modality. We propose to overcome this limitation by combining views from different echocardiographic windows. Following recent publication of a method to register apical and parasternal acquisitions, here we present an algorithm to estimate cardiac motion in RT3DE sequences, using images acquired from both views. The algorithm is based on optical flow, using a variational formulation and a similarity function based on ultrasound physics. Results on simulated and real images are presented, showing the increased accuracy with respect to the same algorithm applied on a single view.

***Index Terms—* echocardiography, motion estimation, optic flow, multiple views**

1. INTRODUCTION

Real-time, three-dimensional echocardiography (RT3DE) has recently become available in clinical practice. This has the potential to overcome many of the limitations of 2D echocardiography, e.g. the need to calculate parameters that are inherently three-dimensional (ventricular volume, myocardial mass) using a very small number of 2D slices, or the limited reproducibility of the studies. However, image quality is still a major shortcoming of RT3DE. This problem is for instance caused by the presence of anatomical structures that block the path of the ultrasound beams, or the dependence of signal strength on the relative angle between the beam and the normal to the anatomical surface. Combining apical and parasternal views may improve results: as the angle between the beam and the surface normal varies between views, structures difficult to appreciate in one of the views are likely to be clearer in the other. Recently, we have proposed and validated a way to align apical and parasternal RT3DE images [1]. The main idea behind this work is that the combination of views can be exploited to improve results on image analysis tasks such as segmentation and motion estimation. In this paper, we explore this idea and propose a motion estimation algorithm that combines images from apical and parasternal views.

Analysis of myocardial motion in echocardiographic sequences is key to the detection of fundamental pathologies such as ischemic heart disease. Many automatic motion estimation methods have been proposed. However, these are not yet routinely used in clinical practice. Different approaches have been explored, e.g. block matching [2], optical flow methods [3,4] or feature tracking [5]. Comparing the relative merits of these is a complex issue and will not be addressed in this paper. To our knowledge, previously published methods have only used a single view. The new approach proposed here combines apical and parasternal views to overcome the limitations of single view echocardiography.

2. METHODS

2.1. Image registration

A necessary first step in our approach is the alignment of apical and parasternal RT3DE sequences. We use the method we recently proposed in [1], which applies a similarity measure based on local phase and orientation differences. The transformation between sequences is rigid in space and piecewise linear in time.

2.2. Motion estimation

Optical flow algorithms are widely used in many applications, providing some of the best results among motion estimation techniques [6]. However, they have several limitations: a) linearizing the difference function introduces inaccuracy, especially in the presence of large deformations; b) these methods are significantly affected by the aperture problem, and c) we cannot assume intensity invariance in echocardiographic sequences. To address limitations a) and b), we use the techniques found in [6,7], which will be briefly presented below. For c), we propose a different similarity function based on ultrasound physics.

Optical flow methods are based on the idea that the intensity of image points remains constant between consecutive frames. By linearizing this constraint using only the first term of the Taylor series, we get the optical flow equation:

$$I_x u + I_y v + I_z w = -I_t \tag{1}$$

where *(u,v,w)* is the 3D velocity vector. An additional regularization must be used to avoid the well-known aperture problem. It is possible to do this using variational techniques. In this case, the method generally involves minimization of an energy function $E=E_{sim}+\alpha E_{smooth}$, where E_{sim} quantifies the difference between two consecutive images using the optical flow constraint $(I_x u+I_y v+I_z w+I_t)^2$, E_{smooth} is the regularization term that quantifies the smoothness of the deformation field, and α determines the relative weights of the two terms.

The similarity term E_{sim} above is a linearized version of the square sum of differences (SSD). Linearization has the advantage of producing a convex energy function, helping convergence. However, if large velocities are present, significant errors can appear. This can be corrected by resampling the image after the energy minimum has been found, and repeating the registration using the resampled image until convergence [6]. A further improvement of the method, also proposed in [6], consists in calculating the value of E_{sim} using, instead of single voxel intensities, the intensities of a region, similarly to the Lucas-Kanade approach. This introduces some of the desirable characteristics of block matching approaches such as [2].

The modifications mentioned above solve some of the shortcomings traditionally associated to optical flow algorithms. However, in the case of ultrasound images, an additional problem remains: intensity constancy is not a valid assumption. An alternative function, optimized for the multiplicative noise of ultrasound images, was used in [2] within a block-matching algorithm:

$$CD_2 = -(I_1 - I_2) + \log(\exp(2(I_1 - I_2)) + 1) \tag{2}$$

We can introduce this in the variational scheme above by using a nonlinear optical flow constraint. Following the notation in [6], the similarity function becomes

$$E_{sim} = \psi(s^2) = -\sqrt{s^2} + \log\left(\exp\left(2\sqrt{s^2}\right) + 1\right) \tag{3}$$

with $s^2=(I_x u+I_y v+I_z w+I_t)^2$.

As mentioned in the Introduction, combining different echocardiographic views can "fill gaps" in the acquired images and thus has the potential of improving the result of motion estimation. In this paper, we assume that cardiac motion is the same in the apical and parasternal sequences. This is a reasonable assumption as long as the two acquisitions are taken in the same session and the cardiac cycle remains similar [1]. We can then minimize:

$$E = E_{sim,apical} + E_{sim,parasternal} + 2\alpha E_{smooth} \tag{4}$$

where $E_{sim,apical}$ and $E_{sim,parasternal}$ are the similarity measures for apical and parasternal sequences, respectively. These two terms are calculated separately using (3) but share the values of *u,v,w*. Note that the smoothing term is common for both acquisitions, as a single motion field is calculated.

In order to validate the calculated motion field, we use it to follow the endocardial surface along the sequence. We do this by manually tracing this surface in a reference frame, and then applying the calculated motion field between each one of the frames and the manually segmented one. We could perform pair-wise alignments between each frame and the reference; however, when the frames are distant in time the presence of large deformations might thwart convergence to the desired solution. On the other hand, if we align adjacent frames we will only have to solve for small deformations, but as we move away from the reference frame the registration errors accumulate and can become very significant. To avoid both problems, we perform a combination of the two mentioned solutions: for each frame, we use the accumulated result of alignment of adjacent frames to initialize the calculation, and then perform an alignment with the reference frame. The final algorithm is (assuming we use frame 1 as reference):

```
For fr=2:number_of_frames
        mot_local=motion field between frames fr and fr-1
        Calculate new frame fr_aligned by applying the motion
        field mot_local+mot_accum to frame fr
        mot_frame=motion field between frames fr_aligned and
        1 (reference)
        mot_accum = mot_accum + mot_frame
end
```

3. RESULTS AND DISCUSSION

3.1. 2D phantom study

In order to show the behavior of the algorithm when using two views containing "gaps", we produced simulated images of a moving elliptical ring. We generated these images by randomly placing a large number (5e4) of point scatterers within the region, with different reflective properties for the three regions: on, inside or outside the ring. We used a predetermined motion field to move scatterers along the sequence. Finally, 2D images for each frame were obtained using the simulation program Field II [8]. Two views were generated, with a variation of 10° in probe angle. We created gaps by locally reducing scatterer reflectivity, as shown in Fig. 1a-d.

Two versions of the algorithm presented in Section 2, using either a single view of both views as in Eq. (4), were then used to calculate motion in the simulated sequence, and this motion was used to track the ring contours. Figure 1e and f show, on a sample frame, the poor results obtained using view 1 or view 2 individually. In Figure 1 g, the results obtained using the presented algorithm are shown. Errors were measured by calculating the Euclidean distance between the obtained contours and ground truth. We show these in Figure 1h. Errors were averaged for individual rows, from top to bottom, to show local error values.

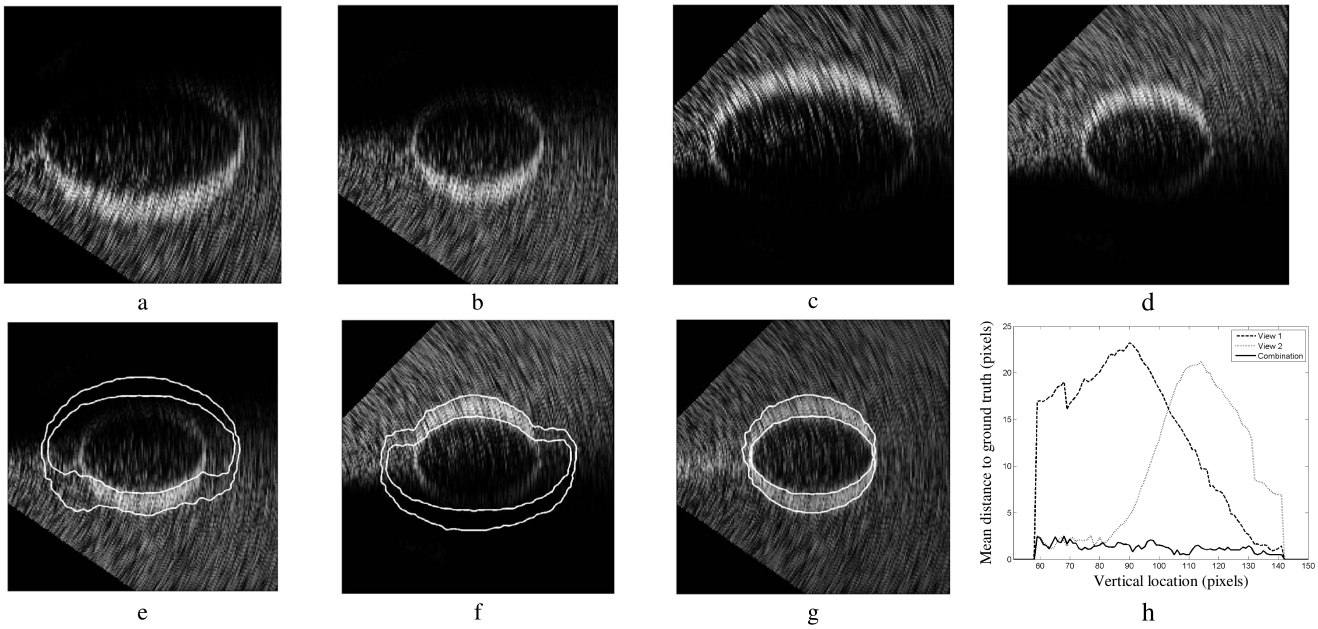

Figure 1. Results on phantom sequences. Top row: sample images from two views of the same object. Images a,b correspond to view 1 at frames 1 and 7, respectively, while c and d correspond to the same frames of view 2. Selective reduction of scatterer reflectivity was performed on the top side of view 1, and the bottom side of view 2. Bottom row: motion estimation results. Panels e and f show the tracked contour for views 1 and 2, respectively. Large errors in the shadowed areas are apparent. Panel g shows the result obtained by combining the two views. For visualization purposes, the contour has been superimposed on the average image of the two views; however, motion estimation is done using the original images and Eq. (4). Finally, in h the errors have been plotted for separate rows from top to bottom of the image.

3.2. RT3DE clinical studies

Though the phantom study provides valuable insight about the behavior of the proposed algorithm, there are obviously many aspects in which these images differ from real echocardiographic images. On the other hand, exact validation of the method on clinical echocardiographic images is complicated as no ground truth is available. We applied the method to 3 pairs of 3D apical/parasternal images, acquired from 3 volunteers. The images were acquired using a Philips Sonos 7500 machine. These were first aligned using the registration method presented in [1], and then motion was calculated using both the single-view and multi-view methods. The endocardial surface was manually segmented in the end-systolic frame, and tracked through the rest of the sequence using the calculated motion field. The whole process was done in 3D. Representative results are shown in Fig. 2. One can see how, in the single-view method using the parasternal view (Fig 2a and e), there are areas in which little information is present (mostly at the apex but also in other locations), and thus the motion field in these regions is very inaccurate. While the apical view offers a much more complete left ventricular boundary, there are also regions where the signal is very weak (see Fig 2b and f for results of the single-view method on the apical view). This is due primarily to the angle between ultrasound beam and surface, as explained in the Introduction. In these areas, the single-view motion estimation algorithm is unable to track the contour, as shown in Fig 2a,b,e,f. The results of the multi-view algorithm, shown in Fig. 2c,d,g,h have a much higher accuracy, particularly at the locations where a single view presents "gaps", highlighted by arrows in Fig. 2.

To obtain quantitative evidence of the improvement provided by the multi-view approach, we manually segmented the end-diastolic frame in one of the apical sequences, and used it as ground truth. Dice coefficients, defined as twice the overlapping volume over the sum of both volumes, were calculated to obtain a measure of agreement between segmentation result and ground truth. Results were obtained using the single-view algorithm with the apical and the parasternal sequence, and the multi-view algorithm combining both. In Figure 3, we show the calculated Dice coefficients; we have calculated them separately on short axis slices to get a better idea of the spatial distribution of the error. It is clear how combining both views significantly improves the result on all slices.

3.3. Discussion

We have presented an algorithm for motion estimation combining several echocardiographic views. Many different algorithms have been proposed to calculate motion estimation, and though some comparison has been done [4], the question about what is the best approach remains open.

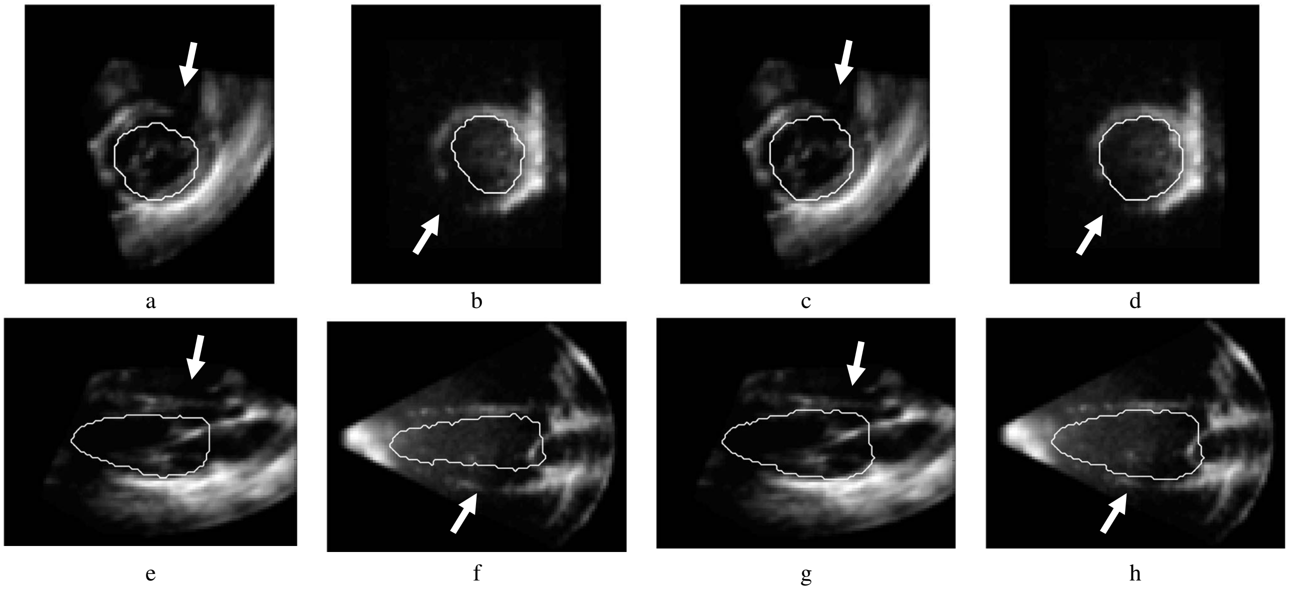

Figure 2. Sample results on a pair of apical / parasternal views. Images a and b show corresponding short axis slices from the parasternal and apical views, respectively. The results of endocardial surface tracking using only one of the views are shown here. The arrows highlight regions in which the motion estimation algorithm has failed. In images c and d, the same slices are shown, now with the contours obtained using the multi-view algorithm. Images e-h show a long axis slice. The shortcomings of the single-view algorithm, and the improvement obtained by using two views, are also apparent.

We have not compared the single-view version of the algorithm presented here with other approaches: the goal of this paper, as mentioned above, is to show an advantage of estimating motion from multiple views in echocardiographic sequences. On the other hand, the method used here, even in the single-view case, has some interesting characteristics, most importantly the use of an ultrasound-specific similarity measure. Though optical flow algorithms can be computationally expensive, with the use of advanced techniques such as multigrid methods it has been shown that results can be obtained in real time for 2D images [7], thus reasonable computation times can be expected in 3D.

We have presented preliminary results on both phantom and clinical data. Before clinical applicability is demonstrated, a larger study will be needed. Using manual segmentation as ground truth is not optimal, as it can be inaccurate and has significant inter- and intraobserver variability. Linking results to clinically relevant parameters such as segment scores, or to MRI, will improve the quality of validation.

Judging by the preliminary results presented here, the main advantage of the multi-view algorithm appears to be in regions with poor image quality. In patients with good echocardiographic windows, the advantage might not compensate the extra cost of acquiring two views. More complete validation is necessary to elucidate this point.

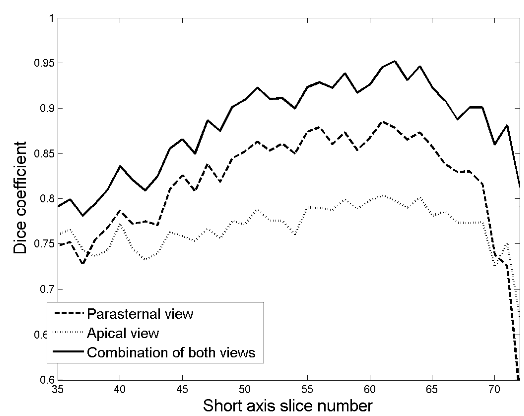

Figure 3. Dice coefficients for single and multiview results

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